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(54) A magnetic resonance imaging method and apparatus

(57) Magnetic resonance is excited **(50)** in first and second species dipoles of a subject in a temporally constant magnetic field. The resonance is refocused **(52)** to generate a spin echo **(54)** centered at a time when the first and second species resonance signals are in-phase. Gradients echoes **(64, 68)** are generated, centered at a time $(2n+1)\pi/\delta\omega$ before and after the spin echo, where $\delta\omega$ is a difference between the first and second species resonance frequencies. In this manner, the first and second species signals are 180° out-of-phase in the gradient echoes. The resonance is refocused **(82)** one or more times to generate additional spin and gradient echoes with different phase encodings **(78)**. The sequence is repeated with yet more phase encodings, and magnetic resonance signals from the spin echo and the two gradient echoes are reconstructed **(86)** into a spin echo image **(s_0)** and a pair of gradient echo images **(s_{+1}, s_{-1})**. A phase map is generated **(90)** from the spin and gradient echo images. One of the gradient echo images is corrected **(116)** with the phase map. The phase corrected gradient image is additively combined **(118)** with the spin echo image to generate a first species image **(112)** and is subtractively combined **(120)** to generate a second species image **(114)**.

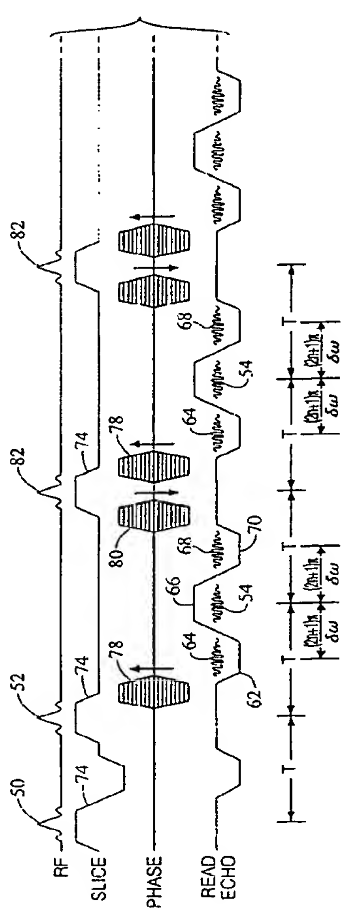

Fig.2

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Description

The present invention relates to a magnetic resonance imaging method and apparatus. It finds particular application in conjunction with the separation of water and fat signals from human patients and will be described with particular reference thereto. However, it is to be appreciated that the present technique is also applicable to imaging sequences in which components of the imaged region have very close resonance frequencies.

Heretofore, subjects have been positioned in a temporally constant magnetic field such that selected dipoles preferentially align with the magnetic field. Radio frequency signals have been applied to cause magnetic resonance of the preferentially aligned dipoles. The radio frequency magnetic resonance signal from the resonating dipoles has been read out for reconstruction into an image representation.

To strengthen the magnetic resonance signal, the resonance can be excited with a 90° radio frequency pulse followed by a 180° refocusing pulse. The 180° refocusing pulse causes the resonating spin system to refocus as a spin echo. The time between the 180° refocusing pulse and the spin echo is the same as the time between the 90° excitation pulse and the 180° refocusing pulse.

Magnetic resonance echoes can also be induced by other disturbances of the spin system, such as reversing the polarity of a magnetic field gradient to induce a gradient echo.

As illustrated in U.S. Patent No. 4,833,407 of Holland, Provost, DeMeester, and Denison, spin and gradient echoes have been induced interleaved in the same repetition of the magnetic resonance imaging sequence. Briefly summarized, an RF excitation and refocusing pulse were applied to induce a spin echo. The polarity of a magnetic field gradient along a read axis was reversed one or more times to induce one or more gradient echoes. When used with fast spin echo (FSE) techniques, refocusing radio frequency pulses are applied after each spin echo to induce yet another spin echo.

As indicated above, water and fat have close resonance frequencies, but differ by about 220 Hertz in a magnetic field of 1.5 Tesla. In the 1.5 Tesla field, the 220 Hertz higher frequency component gains a full revolution on the slower component every 4.46 msec. That is, the signals are in-phase with a 4.46 msec. periodicity. By inducing one echo at 2.23 msec. after refocusing and another echo at 4.46 msec. after refocusing, a pair of echoes can be induced and their signals read out. One of the echo signals has the water and fat in-phase and the other has the water and fat signals 180° out-of-phase. The two echoes can be induced in two different repetitions of the spin echo imaging sequence by shifting the radio frequency refocusing pulse by 2.23 msec. See, for example, Glover, et al., "Three-Point Dixon Technique For True Water/Fat Decomposition with B_0 Inhomogeneity Correction", Magnetic Resonance in Medicine, Vol. 18, pp. 371-383 (1991). One of the problems with this technique is that the system is typically not sufficiently linear that one can add the in-phase and out-of-phase signals to get a water only signal and subtract the in-phase and out-of-phase signals to obtain a fat only signal. The time evolution of the fat and water signals is sufficiently non-linear that in-phase and out-of-phase components do not cancel completely. To correct for this non-linearity, the Glover, et al. technique generates three echoes - one at a nominal echo time, one 2.23 msec. advanced from the nominal echo time, and one 2.23 msec. retarded from the nominal echo time. Through the use of post-processing, Glover seeks to remove the non-linearities such that components of 2.23 msec. displaced echoes substantially cancel. Another drawback of the Glover technique is that three repetitions of the imaging sequence are required to generate magnetic resonance echoes with retarded, advanced, and reference timings.

In accordance with one aspect of the present invention, a magnetic resonance imaging method is provided. A resonance excitation pulse and a resonance refocusing pulse are applied to generate a spin echo. Read gradients are reversed in order to induce first and second gradient echoes in association with each spin echo. The spin echo is induced when the resonance signals of a first species and a second species are substantially in-phase. The gradient echoes are generated at a time when the resonance signals of the first and second species are substantially 180° out-of-phase. In this manner, an in-phase spin echo and out-of-phase gradient echoes are generated in each repetition of the sequence.

In accordance with another aspect of the present invention, a magnetic resonance imaging apparatus is provided. A primary magnet generates a temporally constant magnetic field. A gradient magnetic field coil generates magnetic field gradients across an imaging region. One or more radio frequency coils induces and refocuses magnetic resonance of dipoles of first and second species within the imaging region and receives magnetic resonance signals from the resonating first and second species dipoles in the examination region. The first and second species dipoles have resonance frequencies which differ by a frequency $\gamma\omega$. A timing and control circuit controls the radio frequency and gradient coils (1) to excite magnetic resonance, (2) to refocus the magnetic resonance to induce a spin echo, and (3) to induce gradient echoes centered at times $n\Pi/\gamma\omega$ before and after a center of the spin echo, where n is an odd integer. A reconstruction processor (1) reconstructs magnetic resonance signals from the spin-echo into a spin echo image, (2) reconstructs magnetic resonance signals from the first gradient echo into a first gradient echo image, and (3) reconstructs magnetic resonance signals from the second gradient echo into a second gradient echo image.

One advantage of the present invention is that the reference, advanced, and retarded resonance echo signals are collected in a single acquisition.

Another advantage of the present invention is that data acquisition is accelerated.

Another advantage of the present invention is that it can be used to differentiate among more than two species.

The invention will now be further described, by way of example, with reference to the accompanying drawings in which:-

FIGURE 1 is a diagrammatic illustration of a magnetic resonance imaging device in accordance with the present invention;

FIGURE 2 is a diagrammatic illustration of a magnetic resonance imaging pulse sequence in accordance with the present invention; and,

FIGURE 3 is an alternate embodiment of the magnetic resonance imaging pulse sequence.

With reference to FIGURE 1, a main magnetic field control **10** controls superconducting or resistive magnets **12** such that a substantially uniform, temporally constant magnetic field is created along a z-axis through an examination region **14**. A magnetic resonance echo means applies a series of radio frequency (RF) and magnetic field gradient pulses to invert or excite magnetic spins, induce magnetic resonance, refocus magnetic resonance, manipulate magnetic resonance, spatially and otherwise encode the magnetic resonance, to saturate spins, and the like to generate magnetic resonance imaging and spectroscopy sequences. More specifically, gradient pulse amplifiers **20** apply current pulses to selected ones or pairs of whole body gradient coils **22** to create magnetic field gradients along x, y, and z-axes of the examination region **14**. A digital radio frequency transmitter **24** transmits radio frequency pulses or pulse packets to a whole body RF coil **26** to transmit RF pulses into the examination region. A typical radio frequency pulse is composed of a packet of immediately contiguous pulse segments of short duration which taken together achieve a selected magnetic resonance manipulation. The RF pulses are used to saturate, excite resonance, invert magnetization, refocus resonance, or manipulate resonance in selected portions of the examination region. For whole body applications, the resonance signals are commonly picked up by the whole body RF coil **26**.

For generating images of limited regions of the subject, local coils are commonly placed contiguous to the selected region. For example, an insertable head coil **30** is inserted surrounding a selected brain region at the isocenter of the bore. The insertable head coil optionally includes local gradient coils **32** which receive current pulses from the gradient amplifiers **20** to create magnetic field gradients along x, y, and z-axes in the examination region within the head coil. A local radio frequency coil **34** is used to excite magnetic resonance and receive magnetic resonance signals emanating from the patient's head. Alternatively, a receive-only local coil can be used in conjunction with body-coil transmission. An RF screen **36** blocks the RF signals from the RF head coil from inducing eddy currents in the gradient coils and the surrounding structures.

With continuing reference to FIGURE 1 and further reference to FIGURE 2, a sequence control circuit **40** controls the gradient pulse amplifiers, the digital transmitter, and a digital radio frequency receiver **38** to generate the pulse sequence of FIGURE 2. A timing means **42** controls an Excitation RF pulse generator **44** and a refocusing RF pulse generator **46**. The resonance excitation pulse generator **44** controls the radio frequency transmitter to cause a resonance excitation RF pulse **50**. The refocusing RF pulse generator **46** causes the radio frequency transmitter to generate a refocusing pulse **52** at a time τ after the excitation pulse. The excitation pulse followed by a refocusing pulse causes a spin echo **54** to occur a time 2τ after excitation. The times τ are selected such that the fat and water resonance signals are substantially in-phase at the center of the spin echo **54**. That is, the time 2τ is selected to be an integer multiple of resonance oscillations at both the resonance frequency of water and the resonance frequency of fat, i.e., $2n\Pi/\delta\omega$, where $\delta\omega$ is the frequency difference between the water and fat protons. Preferably, the time τ is also selected such that the fat and water are substantially in-phase when the refocusing pulse **52** is applied.

The timing means **42** also controls a read gradient control **60**. The read gradient control **60** causes a first read gradient **62**. The application of the read gradient **62** is timed such that a gradient echo **64** is induced at a time $(2n+1)\Pi/\delta\omega$ before the center of the spin echo **54**. A center lobe **66** of the read gradient is applied centered on the spin echo **54**. The read gradient control again reverses to induce a second gradient echo **68** at time $(2n+1)\Pi/\delta\omega$ after the spin echo and concurrently with a read gradient lobe **70**. In this manner, a spin echo is generated surrounded on either side by a pair of gradient echoes which are 180° or Π out-of-phase with the spin echo. Again, for fat and water in a 1.5 Tesla field, $\Pi/\delta\omega=2.23$ msec.

The timing means also controls a slice select gradient control **72** which controls the gradient amplifiers **20** to cause slice select gradient pulses **74** concurrently with the RF excitation and refocusing pulses. The slice select gradients limit the examined region to a single slice. Alternately, slab select pulses may be applied and phase encode pulses can be applied to encode the resultant spin and gradient echoes along the slice select axis. The timing means further controls a phase encode gradient controller **76** which causes the gradient amplifiers **20** to induce phase encode gradient pulses **78** before each of the spin and gradient echoes. The phase encode gradient echoes are applied such that the spin echo and its two associated gradient echoes have the

same phase encoding. Preferably, a dephasing gradient pulse **80** can be applied along the phase encode axis to remove or zero the phase encoding after the last gradient echo.

In the preferred embodiment, the radio frequency refocusing pulse controller **46** causes subsequent refocusing pulses **82** at a time 2τ after the first refocusing pulse and at 2τ intervals thereafter. The read, slice, and phase encode gradient controllers cause like gradient pulses following each refocusing pulse to collect a second and subsequent sets of image data.

With reference again to FIGURE 1, the magnetic resonance signals from the spin echo and the pair of gradient echoes are received by the whole body RF coil **26** or the localized coil **34** and conveyed to the digital receiver **38**. A sorter **84** sorts the signals from the reference, retarded, and advanced echoes. A reconstruction processor **86**, preferably three parallel processors, reconstructs a reference image s_0, a first, retarded image s_{-1}, and a second, advanced image s_{+1}.
The images are defined by:

$$s_{-1} = (\rho_W - \rho_F)\, e^{i(-\phi+\phi_{rf})} \qquad (1a),$$

$$s_0 = (\rho_W + \rho_F)\, e^{i(\phi_{rf})} \qquad (1b),$$

$$s_{+1} = (\rho_W - \rho_F)\, e^{i(+\phi+\phi_{rf})} \qquad (1c),$$

where ρ_W and ρ_F denote the real water and fat components, respectively, ϕ represents a complex phase to the local field error (also known as background phase), and ϕ_{rf} is the systematic phase error introduced from the RF hardware chain. The s_{-1}, s_0, and s_{+1} images are stored in image memories **88**$_-$, **88**$_0$, **88**$_+$, respectively.

A phase map or error generator **90** analyzes the images s_{-1}, s_0, and s_{+1} to determine a map of the phase error at each pixel of an imaged slice. For calculational simplicity, a transform circuit **92** generates complex phase signals S_+ and S_- defined by:

$$S_+ = \frac{s_{+1}+s_{-1}}{2s_0} = \frac{(\rho_W-\rho_F)}{(\rho_W+\rho_F)} \cos\phi \qquad (2a),$$

$$S_- = \frac{s_{+1}-s_{-1}}{2s_0} = \frac{(\rho_W-\rho_F)}{(\rho_W+\rho_F)}\, i \sin\phi \qquad (2b).$$

The complex phase due to the local field error, i.e., the background phase, is given by:

$$\mathrm{Arg}(\mathrm{Re}(S_+)+ i\, \mathrm{Im}(S_-)) = \phi \pm 2\Pi n - \Pi p \qquad (3),$$

where n denotes an integer and p varies between 0 and 1. A phase map processor **94** processes the output of the transform **92** to generate a phase map in accordance with Equation (3) for storage in a phase map memory **94**. Of course, the phase map has both 2Π and Π discontinuities. An unwrapping algorithm or processor **96** unwraps or adjusts the phase map to eliminate the Π and 2Π discontinuities to recover the background phase. In the preferred embodiment, the background phase is fit to a polynomial ϕ_{fit} using a least squares processing routine **98**. In the preferred embodiment in which the background phase is fit to a polynomial, it is defined by:

$$\phi_{fit} = \phi_0 + a_1x + b_1y + a_2x^2 + b_2y^2 + c_2xy + \ldots \qquad (4).$$

The background phase $e^{\phi_{fit}}$ for each pixel is stored in a phase correction or background phase memory **100**.

A corrected image generator **110** combines the phase correction and the uncorrected reconstructed images to generate phase corrected water and fat images which are stored in a water image memory **112** and a fat image memory **114**. In the preferred embodiment, the water and fat images are defined by:

$$\text{water image} = \left| s_0 + s_{-1} e^{i\phi_{fit}} \right| \qquad (5a),$$

$$\text{fat image} = \left| s_0 - s_{-1} e^{i\phi_{fit}} \right| \qquad (5b).$$

More specifically, a multiplier **116** multiplies one of the s_{-1} and s_{+1} images with the phase correction from the background phase memory **100**. An image adder **118** adds the complex phase corrected gradient echo image with the complex spin echo image to generate the water image (Equation (5a)). A subtraction circuit **120** subtractively combines the complex phase corrected gradient echo with the complex spin echo image to generate the fat image (Equation (5b)). Optionally, a weighting adjustment **122** is provided for multiplying the phase adjusted complex gradient echo image by an adjustable weighting factor to adjust the relative weighting between the spin echo image and the gradient echo image.

A video processor **124** converts selectable slices or other portions of the water or fat images or combinations of the two into appropriate format for display on a monitor **126** or other human readable display device.

Of course, this technique can also be used to distinguish between other than fat and water. With reference to FIGURE 3, the read gradient can have five lobes for distinguishing among three species. The values of τ are selected such that the spin echo **54** occurs when all three species are in-phase. A first pair of gradient echoes **64, 68** for a first species are generated at a time $\Pi/\delta\omega'$ before and after the spin echo, where $\delta\omega'$ is the fre-

quency difference between the resonance frequency of the second species and the first species. A second pair of gradient echoes **130, 132** are generated a time $n\Pi/\delta\omega'$ before and after the spin echo, where $\delta\omega'$ is the frequency difference between the first and third species and n is an integer, preferably 1 or an odd number. The above-described phase correction process is repeated to generate a phase correction between the first and second species and another phase correction between the first and third species. The sum of the complex gradient echo image that is adjacent to the spin echo and the complex spin echo image itself provides an image of the first species. The sum of the complex gradient echo images that are farthest from the spin echo and the complex spin echo image itself provide an image of the first species, for example, water. The subtraction of the same sets of images provide images of the second and third species, for example, fat and silicone, respectively.

Claims

1. A magnetic resonance imaging method in which a resonance excitation pulse and a resonance refocusing pulse are applied to generate a spin echo (54) and in which read gradients are reversed in order to induce first and second gradient echoes (64, 68) in association with each spin echo, characterized by:

   inducing the spin echo (54) when resonance signals of a first and second species are substantially in-phase; and,
   generating the gradient echoes (64, 68) at a time when the resonance signals of the first and second species are substantially 180° out-of-phase such that an in-phase spin echo and out-of-phase gradient echoes are generated in each repetition of the imaging sequence.

2. A method as claimed in claim 1, including:

   reconstructing a spin echo image from the spin echo signals from a plurality of repetitions of the imaging sequence;
   generating first and second gradient echo images from the gradient echo signals of the plurality of repetitions of the imaging sequence;
   generating a phase correction map from the spin echo image and the first and second gradient echo images;
   correcting phase error in the first gradient echo image in accordance with the phase map;
   additively combining the spin echo image with the phase corrected first gradient echo image to generate a first species image (112); and
   subtractively combining the spin echo image with the phase corrected first gradient echo image to generate a second species image (114).

3. A method as claimed in claim 2, further including:

   generating the first gradient echo (64) a time $\Pi/\delta\omega$ before the spin echo, where $\delta\omega$ is a difference in the resonance frequencies of the first and second species; and,
   generating the second gradient echo (68) at a time $\Pi/\delta\omega$ after the spin echo.

4. A method as claimed in claim 3, further including:

   generating the spin echo (54) at a time when resonance signals of the first species, second species, and a third species are all in-phase;
   generating a third gradient echo (130) at a time $\Pi/\delta\omega'$ after the spin echo, where $\delta\omega'$ is a difference in the resonance frequencies between the first and third species; and
   generating a fourth gradient echo (132) at a time $\Pi/\delta\omega'$ before the spin echo.

5. A method as claimed in any of the preceding claims, wherein the first species (112) is water and the second species (114) is fat.

6. A method as claimed in any of the preceding claims, further including:

   applying an additional refocusing RF pulse after the spin echo and the first and second gradient echoes to induce a second spin echo at a time when the first and second species are again in-phase; and,
   generating a pair of additional gradient echoes after the second refocusing pulse at times when the first and second species are 180° out-of-phase.

7. A method as claimed in any of the preceding claims, further including phase encoding the spin and gradient echoes with a common phase encoding in each of a plurality of repetitions.

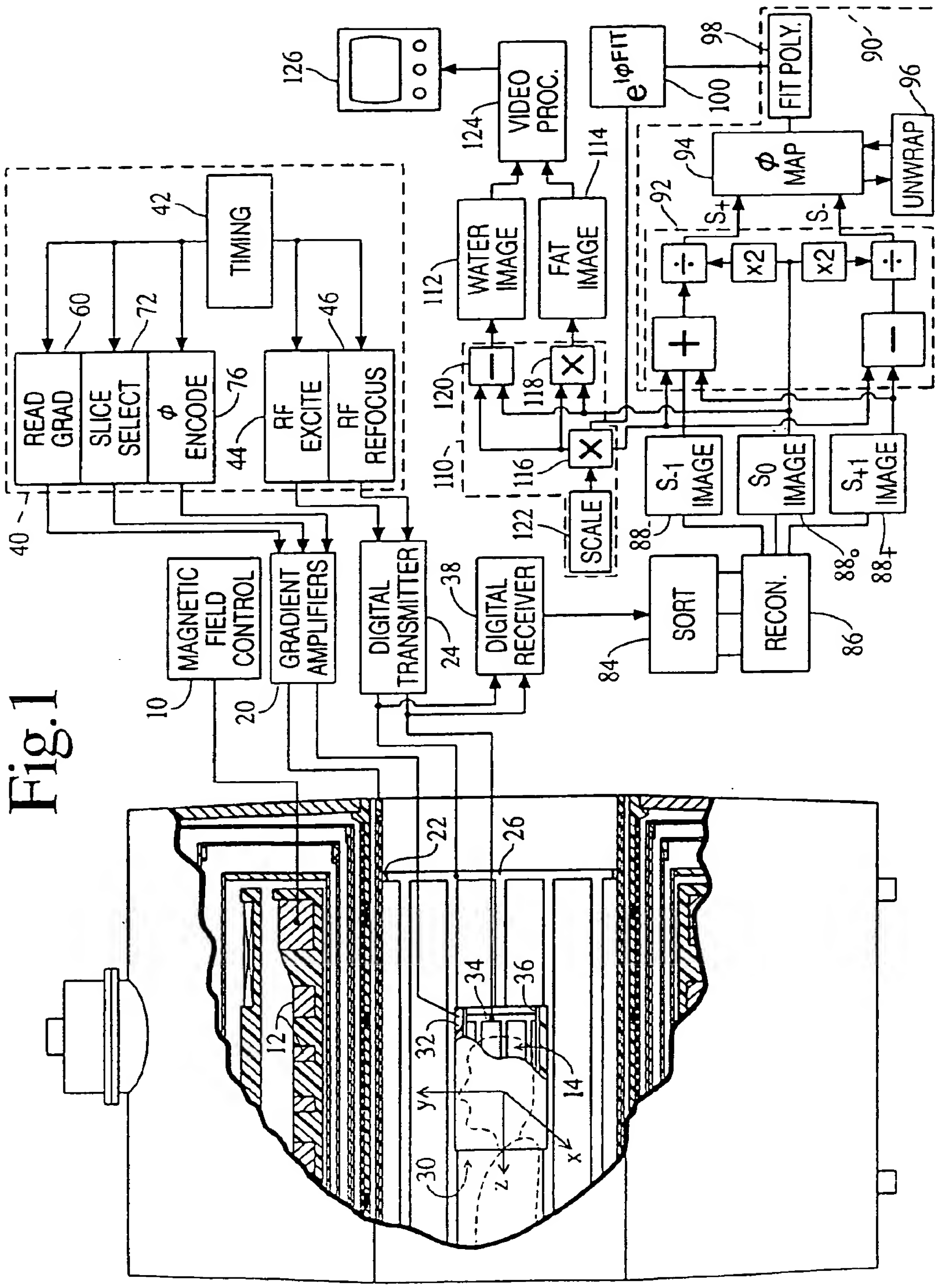
8. A magnetic resonance imaging apparatus which includes a primary magnet (12) for generating a temporally constant magnetic field, a gradient magnetic field coil (22) for generating magnetic field gradients across an imaging region, one or more radio frequency coils (26) for inducing and refocusing magnetic resonance of dipoles of first and second species within the imaging region and for receiving magnetic resonance signals from the resonating first and second species dipoles in the examination region, the first and second species dipoles having resonance frequencies which differ by a frequency

difference $\delta\omega$, characterized by:

a timing and control circuit (40, 42) for controlling the radio frequency and gradient coils to (1) excite magnetic resonance, (2) refocus the magnetic resonance to induce a spin echo (54), and (3) induce gradient echoes (64, 68) centered at times $n\Pi/\delta\omega$ before and after a center of the spin echo, where n is an odd integer;
a reconstruction processor (86) for (1) reconstructing magnetic resonance signals from the spin echo into a spin echo image, (2) reconstructing magnetic resonance signals from the first gradient echo into a first gradient echo image, and (3) reconstructing magnetic resonance signals from the second gradient echo into a second gradient echo image.

9. A magnetic resonance imaging apparatus as claimed in claim 8, further including:

a phase map generator (90) which generates a phase correction map from the spin echo image and the first and second gradient echo images;
a circuit (110, 116) which corrects one of the gradient echo images with the phase map to generate a phase map corrected gradient echo image;
an image adder (118) which adds the spin echo and the phase corrected gradient echo images to generate a first species dipole image (112);
an image subtractor (120) which subtractively combines the spin echo and the phase corrected gradient echo images to generate a second species dipole image (114).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RF EXCITE
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DIGITAL RECEIVER
SORT
RECON.
SCALE
S_{-1} IMAGE
S_0 IMAGE
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FAT IMAGE
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Fig.1

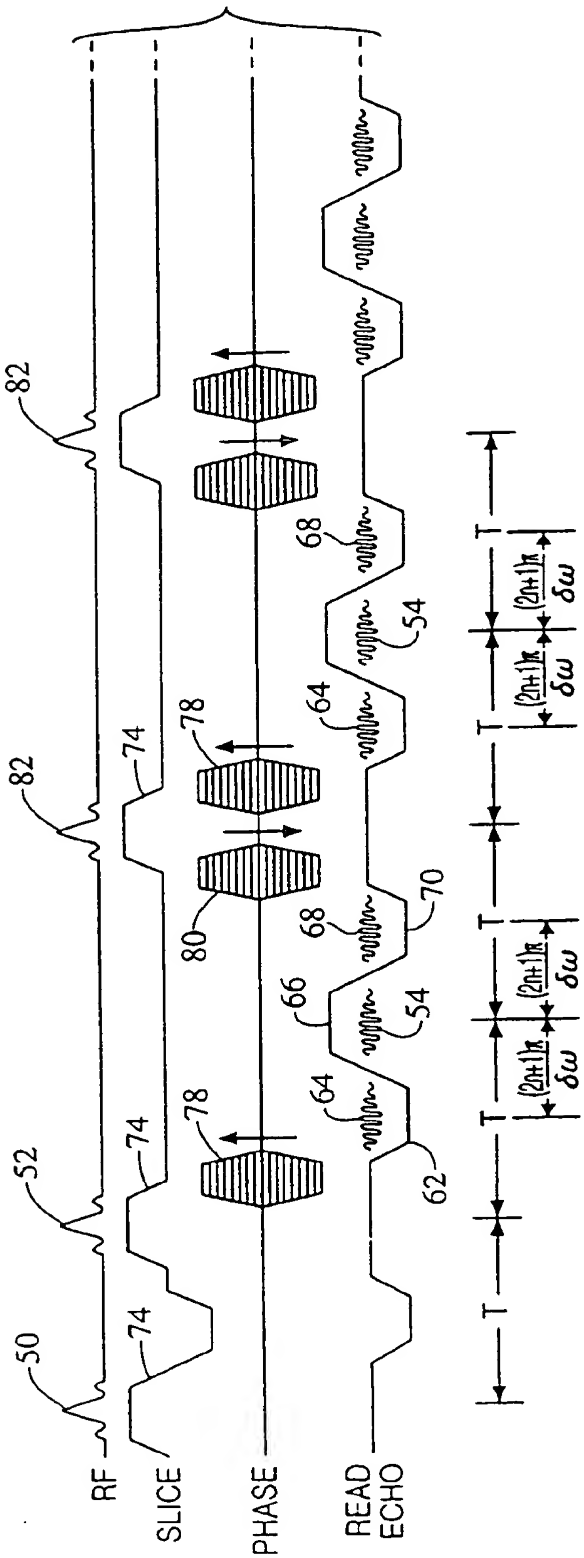
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Fig.2

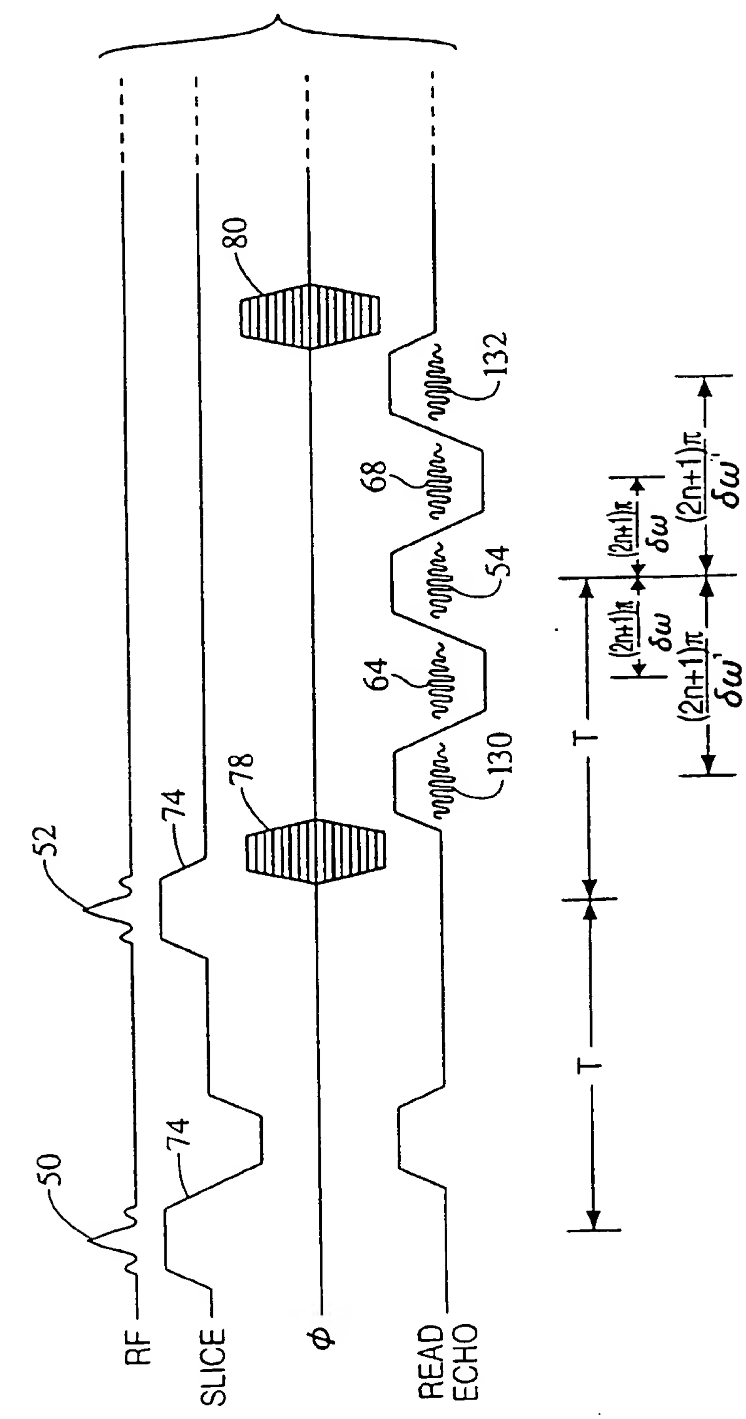

Fig.3

European Patent Office

EUROPEAN SEARCH REPORT

Application Number
EP 96 30 3300

DOCUMENTS CONSIDERED TO BE RELEVANT			
Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int.Cl.6)
D,A	MAGNETIC RESONANCE IN MEDICINE, vol. 18, 1991, pages 371-383, XP000209847 G.H. GLOVER, E. SCHNEIDER: "Three-Point Dixon Technique for True Water/Fat Decomposition with B0 Inhomogeneity Correction" * pages 371 - 377: paragraphs I and II * * pages 379 and 380: paragraph IV *	1-9	G01R33/485
A	US-A-5 051 699 (M. HANAWA, K. NAKABAYASHI) * column 1, line 6 - line 20 * * column 2, line 61 - column 3, line 24 * * column 4, line 26 - line 68 * * column 6, line 40 - column 7, line 39 * * figures 3,6,7 *	1-3,5,7,8	
A	US-A-5 250 899 (J. LISTERUD, T. CHAN) * column 3, line 21 - column 4, line 52; figure 1 *	1,8	TECHNICAL FIELDS SEARCHED (Int.Cl.6)
A	EP-A-0 417 284 (YOKOGAWA MEDICAL SYSTEMS, LTD) * page 4, line 46 - page 6, line 29; figure 1 *	1-3,5,7-9	G01R

The present search report has been drawn up for all claims

Place of search	Date of completion of the search	Examiner
THE HAGUE	11 September 1996	Volmer, W

CATEGORY OF CITED DOCUMENTS

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